# Contribution of Knee Flexor and Extensor Strength on Sex-Specific Energy Absorption and Torsional Joint Stiffness During Drop Jumping

By: Randy J. Schmitz and Sandra J. Shultz

Schmitz RJ and Shultz SJ. Contribution of Knee Flexor/Extensor Strength on Sex-Specific Energy Absorption and Torsional Joint Stiffness Patterns during Drop Jumping. *Journal of Athletic Training* 2010; 45(5):445-52.



## Abstract:

**Context:** Lower extremity injury often occurs during abrupt deceleration when attempting to change the body's direction. Although sex-specific biomechanics have been implicated in the greater risk of acute knee injury in women than in men, it is unknown if sex differences in thigh strength affect sex-specific energy absorption and torsional joint stiffness patterns.

**Objective:** To determine sex differences in energy absorption patterns and joint stiffnesses of the lower extremity during a drop jump and to determine if these sex differences were predicted by knee extensor and flexor strength.

**Design:** Cross-sectional study.

**Setting:** Laboratory environment.

**Patients or Other Participants:** Recreationally active, college-aged students (41 women: age = 22.1 ± 2.9 years, height = 1.63 ± 0.07 m, mass = 59.3 ± 8.0 kg; 40 men: age = 22.4 ± 2.8 years, height = 1.77 ± 0.1 m, mass = 80.9 ± 14.1 kg).

**Intervention(s):** Participants performed knee flexor and extensor maximal voluntary isometric contractions followed by double-leg drop-jump landings.

**Main Outcome Measure(s):** Lower extremity joint energetics ($J \times N^{-1} \times m^{-1}$) and torsional joint stiffnesses ($Nm \times N^{-1} \times m^{-1} \times degrees^{-1}$) were calculated for the hip, knee, and ankle during the initial landing phase. Body weight was measured in newtons and height was measured in meters. Sex comparisons were made and sex-specific regressions determined if thigh muscle strength (Nm/kg) predicted sagittal-plane landing energetics and stiffnesses.

**Results:** Women absorbed 69% more knee energy and had 36% less hip torsional stiffness than men. In women, greater knee extensor strength predicted greater knee energy absorption ($R^2$ = 0.11, $P$ = .04), and greater knee flexor strength predicted greater hip torsional stiffness ($R^2$ = 0.12, $P$ = .03).

**Conclusions:** Sex-specific biomechanics during the deceleration phase of a drop jump revealed that women used a strategy to attempt to decrease system stiffness. Additionally, only female strength values were predictive of landing energetics and stiffnesses. These findings collectively demonstrated that the task may have been more difficult for women, resulting in a different movement strategy among those with different levels of thigh strength to safely complete the task. Future researchers should look at other predictive factors of observed sex differences.

**Keywords:** work | energetics | lower extremity biomechanics

**Article:**

**Key Points**

- During drop jumping, women absorbed more energy about the knee than did men.
- Men had greater hip stiffness than women.
- Greater knee extensor strength predicted greater knee energy absorption during drop jumping in women.
- Greater knee flexor strength predicted greater hip torsional stiffness during drop jumping in women.
- A large proportion of the variance in lower extremity mechanics during drop jumping was not explained by thigh strength.

Internal and external torques associated with lower extremity joint function are likely affected by differential kinematic patterns of lower extremity joint function.[1,2] Researchers[3] have suggested that these kinematic patterns are modulated by muscular activation patterns. Work done on the extensor muscles through eccentric muscle action during landing can be described as *energy* absorption.[4] Furthermore, manipulation of the entire lower extremity, such as an alteration in leg spring stiffness, is greatly influenced by contributions of individual torsional joint stiffnesses.[5,6]

Researchers have investigated voluntary, cognitive manipulations of stiffness levels during landing and how they affect energy absorption patterns about the hip, knee, and ankle during landings. They[1,2] reported that increases in stiffness (as assessed through a reduction in joint flexion range of motion) were associated with landings characterized by less energy absorption. They suggested that extensor muscle control was responsible for these differential effects, as a higher level of extensor activity could result in greater stiffness and less energy absorption. However, we do not know whether the inherent maximal strength-producing capabilities of the muscles may affect torsional joint stiffness (change in moment/change in position[5] during the energy absorption phase of an elastic motion) and the corresponding energy absorption during impulse application.

Understanding the relationship between maximal strength capabilities and torsional stiffness may be clinically important, as increasing or decreasing torsional stiffness of the lower extremity is associated with increasing or decreasing overall system stiffness and potentially affecting injury risk.[5] In a recent review, investigators[3] suggested that increasing system stiffness may be associated with bony injuries, such as stress fractures and osteoarthritis, as increased stiffness is typically associated with a reduction in the range of motion used and greater peak forces. Conversely, decreasing system stiffness has been associated with excessive joint motion, leading to soft tissue injury.[3,7] Thus, it may be important to include torsional stiffness measures in biomechanics investigations centered on understanding injury mechanisms and risk factors, particularly as they relate to potential underlying causes for sex differences in knee ligament injuries.

Biomechanical investigations of movement patterns during decelerating landing activities involve terminal landing tasks (eg, single-leg or double-leg drop landings) and nonterminal landing tasks (eg, countermovement jumps or drop jumps). In a report of sex differences in landing energetics, Decker et al[8] demonstrated that female recreational athletes had more erect landing posture and greater energy absorption from the knee extensors and ankle plantar flexors during terminal double-leg drop landings; however, Schmitz et al[9] demonstrated that women had less total lower body energy absorption and greater relative energy absorption at the ankle than did men during a terminal single-leg landing. The different demands (stabilization of the system using 2 limbs or 1 limb, respectively) of the task likely accounted for the differential results. We do not know if such sex differences would exist in a nonterminal activity, such as a drop jump.

The method of quantifying energy absorption by the extensor muscles is based on the integration of the negative power curve.[4] A negative power curve is generated when the direction of joint motion is in opposition to the internal moment generated, which can functionally be interpreted as an eccentric muscle action. This carries an assumption that underlying moment-producing capability would affect this measure. However, we were unable to locate sex comparison studies in which predictive factors of energy absorption or torsional stiffness were investigated. In a study of energy absorption during a terminal landing task in gymnasts and recreational athletes, McNitt-Gray[4] suggested that the lesser extensor moments produced by recreational athletes during higher-demand tasks may have been the source of their greater hip flexion excursion and longer phase duration of landings. However, they found no differences in lower extremity energy absorption between the groups. Although strength was not directly assessed, one could infer that absolute strength differences or use of available strength may have played a role.

Using computer modeling, Sandler and Robinovitch[10] demonstrated that strength plays a factor in attenuating the body's vertical kinetic energy from a fall. Specifically, decreased strength adversely affects energy absorption mechanisms. Because increasing strength is thought to be a necessary component of programming for the prevention of anterior cruciate ligament (ACL) injury,[11] it is conceivable that lesser strength (common in female-to-male comparisons) may affect the ability of athletes to perform less risky landing mechanics.[12] Because the inability to control impact absorption may lead to musculoskeletal injury,[13] researchers have suggested that a better understanding of energy absorption pattern differences between sexes may lend insight into the mechanisms behind the greater risk of ACL injury in females.[8]

Many researchers[1,2,8,9] have investigated energy absorption patterns during terminal drop landings, but few researchers[14–16] have studied energy absorption patterns of the negative or braking phase of nonterminal landing tasks. Because many acute joint injuries to the lower extremity occur when the body is attempting to change its direction of motion (ie, cutting or countermovement-jump task),[17] examining the initial landing or braking phase of a vertical drop jump offers an established and relevant model to study this sport-specific maneuver.[18] We could not identify any studies in which sex-specific energy absorption patterns during these nonterminal landing tasks were investigated. Additionally, we could not locate an in vivo study in which the relationship of muscle strength to energy absorption patterns and torsional joint stiffnesses was studied. Thus, the purposes of our study were (1) to determine sex differences in energy absorption patterns and joint stiffnesses of the lower extremity during a drop jump and (2) to determine if these sex differences were predicted by knee extensor and flexor strength.

## METHODS

### Design

As part of a larger study that we conducted, participants attended 1 testing session in which all measures for this study were obtained. For women, testing was performed during the first 6 days of menses to control for any potential acute hormonal effects on neuromuscular control.[18] During the session, participants completed maximal voluntary isometric contraction (MVIC) strength assessments, which have been reported[18] for a subset of the included population. This was followed by 5 double-leg drop-jump landings. All testing was performed on the dominant-stance limb (the leg on which the participant would stand when kicking a ball).

### Participants

Eighty-one physically active, healthy, college-aged students (41 women: age = 22.1 ± 2.9 years, height = 1.63 ± 0.07 m, mass = 59.3 ± 8.0 kg; 40 men: age = 22.4 ± 2.8 years, height = 1.77 ± 0.1 m, mass = 80.9 ± 14.1 kg) volunteered to participate in this study. *Physically active* was defined as engaging in 2 to 10 hours of physical activity each week. *Healthy* was defined as having no previous orthopaedic injury or neurologic disorder of the lower extremity that impaired performance during recreational activity. These data were obtained through a medical history and activity questionnaire. Participants provided written informed consent, and the study was approved by the university's institutional review board. Participants were familiarized with all testing procedures approximately 2 weeks before testing.[18]

### Instrumentation

Kinematic data for the foot, shank, thigh, pelvis, and trunk of the dominant limb were collected at 100 Hz using an electromagnetic tracking system (Motion Star; Ascension Technologies, Burlington, VT). Six–degrees-of-freedom position sensors were attached to each participant's dominant limb with double-sided tape and elastic wrap over the anterior mid-shaft of the third metatarsal, the mid-shaft of the medial tibia, and the lateral aspect of the mid-shaft of the femur. Sensors were also placed on the contralateral femur and tibia, on the sacrum, and over the C7 spinous process. Vertical ground reaction force data were obtained at 1000 Hz with a Bertec force plate (model 4060-NC; Bertec Corporation, Columbus, OH).

Before collection of the landing trials, MVICs of the knee extensors and knee flexors were collected. Participants were positioned in a Biodex System 3 isokinetic dynamometer (Biodex Medical Systems Inc, Shirley, NY) at 20° of knee flexion and were instructed to complete three 5-second maximal isometric quadriceps and hamstrings contractions (MVICs).

## Vertical Drop-Jump Protocol

Participants performed bilateral, barefooted, drop-jump landings from a wooden platform measuring 0.45 m in height and placed 0.1 m behind the rear edge of the landing target (force plate).[18] They were barefooted to control for the effect of varied footwear across participants. For all landings, participants began in a standardized take-off position in which the hands were held just lateral to the shoulders with the palms facing anteriorly and the toes of the dominant foot were aligned along the leading edge of the wooden platform.[18] Participants were instructed to jump down; land simultaneously on both feet (dominant foot centered on the force plate and nondominant foot completely off the plate); and complete a rapid, maximal, double-leg jump effort. To prevent experimenter bias, we did not give special instructions to participants regarding their landing mechanics.[18] Practice repetitions were performed until the participant appeared to be comfortable and reported comfort with the task (typically 3 to 6 repetitions) to reduce the potential for learning effects. Five jumps were recorded for analyses. The data interval from initial foot contact on the force plate to the point at which the body's center of mass reached its lowest position before beginning the propulsive phase of the double-leg jump was used for the analyses.

## Data Processing

Ground reaction force data were offline, low-pass filtered at 60 Hz using a fourth-order, zero-lag Butterworth filter. The *landing phase* was defined as the time from the point at which vertical ground reaction force exceeded 10 N to the time at which the body's center of mass reached its lowest position. Peak vertical ground reaction forces were normalized to body weight (N).

Hip joint centers were calculated using the method of Leardini et al.[19] Knee joint centers were calculated as the centroid of the medial and lateral femoral epicondyles, and ankle joint centers were calculated as the centroid of the medial and lateral malleoli.[20] A segmental reference system was defined for all body segments, with the *positive z-axis* defined as the left-to-right axis, the *positive y-axis* defined as the distal-to-proximal longitudinal axis, and the *positive x-axis* defined as the posterior-to-anterior axis. Three-dimensional hip, knee, and ankle flexion angles were calculated using Euler angle definitions with a rotational sequence of $Z\ Y'\ X''$.[21] Raw kinematic data were linearly interpolated to force-plate data and were subsequently low-pass filtered at 12 Hz using a fourth-order, zero-lag Butterworth filter.

*Total flexion displacement* for the hip, knee, and ankle was defined as the difference between the joint angle at ground contact and the peak joint angle. For consistency, increasing hip, knee, and ankle flexion are reported as positive values. Hip, knee, and ankle intersegmental moments were calculated using an inverse dynamics analysis within The MotionMonitor software (Innovative Sports Training, Chicago, IL).[22] All angular kinetics were normalized to the product of body weight (N) and height (m). Net joint powers were calculated as the product of the normalized joint moment and joint angular velocity (the derivative of the angular position calculated) at each time point. Next, work done on the extensor muscles was calculated by integrating the negative portion of the joint power curve, as this represented energy absorption by the extensor muscles.[4] Thus, work values are reported as normalized to the product of body weight and height ($J \times N^{-1} \times m^{-1}$). For consistency purposes, energy absorption of the hip, knee, and ankle extensors are reported as positive values. Sagittal-plane hip, knee, and ankle torsional joint stiffnesses were calculated as the change in normalized net internal moment (Nm) divided by the change in angular position (degrees) from initial contact to peak flexion excursion ($Nm \times N^{-1} \times m^{-1} \times degrees^{-1}$) during the defined landing phase.[5]

## Statistical Analyses

To validate our measures of torsional joint stiffness, averaged moment-joint displacement curves were calculated from individual trial data that were normalized to 101 points for the defined landing phase. From these plots, we performed linear regressions to help determine the degree of linear relationship between the 2 variables. Strong linear relationships (high $R^2$ values) would indicate that the system behaved like a spring-mass model, helping to validate the torsional stiffness model for use in this specific task.

The averages of the multiple trials were used for all dependent variables. We used a mixed-factor, repeated-measures analysis of variance (ANOVA) (sex × joint) to test for sex differences in energy absorption and torsional joint stiffness. We used post hoc Tukey honestly significant difference testing to test the omnibus $F$ value. Sex differences in strength were assessed through 1-way ANOVAs. We used stepwise linear regression analysis ($P = .51$ removal using probability of $F$ stepping method criteria in SPSS [version 16.0; SPSS Inc, Chicago, IL]) to examine the extent to which thigh strength (knee flexor MVIC and knee extensor MVIC) was associated with energy absorption and torsional joint stiffnesses (dependent variables) within each sex and joint. The α level was set at $P = .05$. We used SPSS for all statistical analyses.

# RESULTS

Averaged extensor moment-sagittal joint displacement curves and resultant linear regression lines are shown in Figures 1 through 3. Linear regression equations for each joint demonstrated highly linear moment-displacement relationships for the hip ($R^2 = 0.854$), knee ($R^2 = 0.90$), and ankle ($R^2 = 0.84$). Thus, we believe that the measure of torsional joint stiffness presented is a valid means to describe the function of individual joints as a spring-mass model system during the initial landing phase of the drop-jump task. In addition, to help with interpretation of the data, we have included sex-specific kinematic curves in Figures 4 through 6; the excursion data have been reported[18] in a subset of the population used in this study.

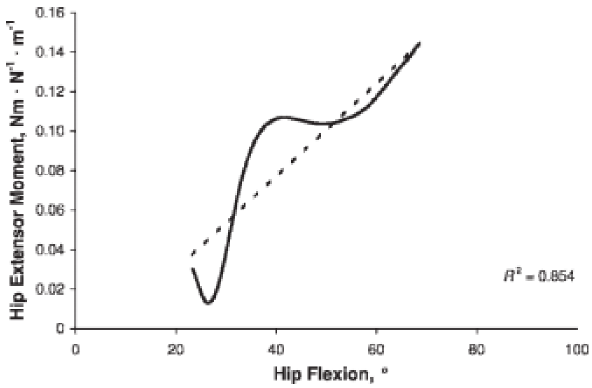


**Figure 1. Averaged hip extensor moment. Body weight was measured in newtons, and height was measured in meters.**

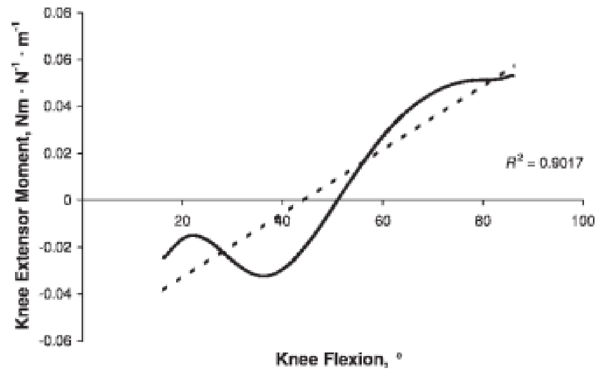


**Figure 2. Averaged knee extensor moment. Body weight was measured in newtons, and height was measured in meters.**

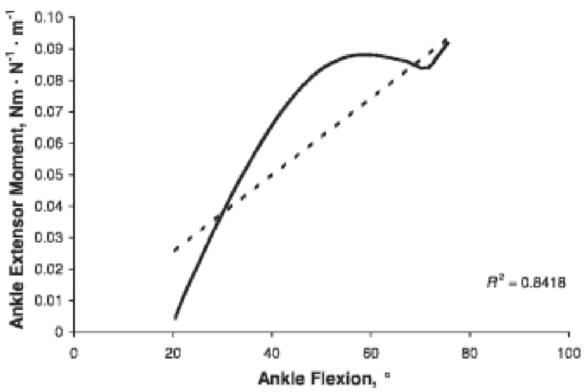


**Figure 3. Averaged ankle extensor moment. Body weight was measured in newtons, and height was measured in meters.**

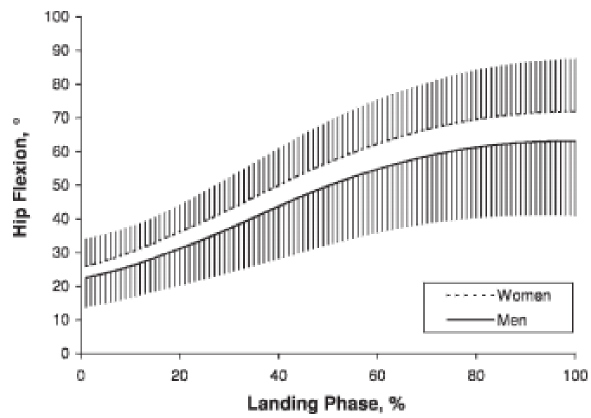


**Figure 4. Mean ± SD for female and male hip flexion during the landing phase.[18]**

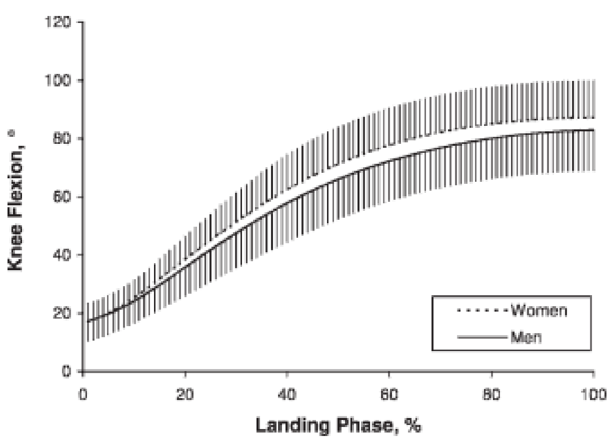


**Figure 5. Mean ± SD for female and male knee flexion during the landing phase.[18]**

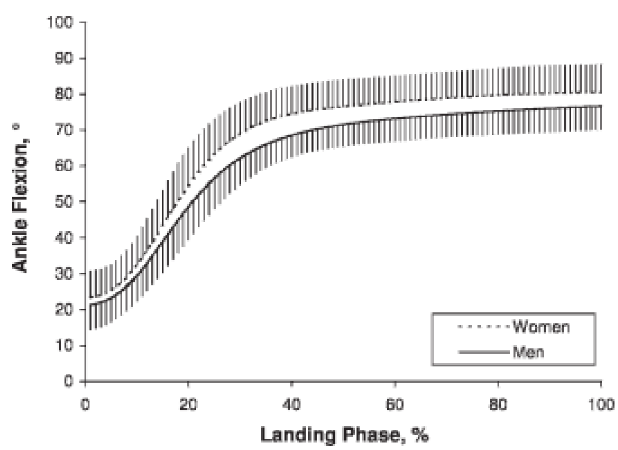


**Figure 6. Mean ± SD for female and male ankle flexion during the landing phase.**

The mean ± SD values, effect sizes, and 95% confidence intervals for sagittal hip, knee, and ankle energy absorption and torsional joint stiffness are reported in the Table. The energy absorption ANOVA revealed an interaction of sex and joint ($F_{2,158}$ = 4.102, $P$ = .02). Post hoc testing revealed that men absorbed less normalized energy than women at the knee joint ($P$ < .05) and demonstrated no differences between sexes at the other joints ($P$ > .05). We did not find a main effect for sex ($F_{1,79}$ = 2.098, $P$ = .29), indicating no sex difference in total hip, knee, and ankle energy absorption. To help with functional interpretation, we secondarily expressed the percentage of total energy absorption for each joint relative to summed hip, knee, and ankle energy absorption (Figure 7). Follow-up 1-way ANOVAs revealed that percentage of knee work absorption was greater in women than in men ($F_{1,79}$ = 14.49, $P$ < .001) but revealed no differences in the hip ($F_{1,79}$ = 1.838, $P$ = .18) or ankle ($F_{1,79}$ = 3.546, $P$ = .06); these values are similar to the absolute values we reported. The joint stiffness ANOVA also revealed an interaction of sex and joint ($F_{2,158}$ = 6.002, $P$ = .003). Post hoc testing showed that men had greater sagittal hip torsional stiffness than women ($P$ < .05) but revealed no differences between sexes at the other joints ($P$ > .05).

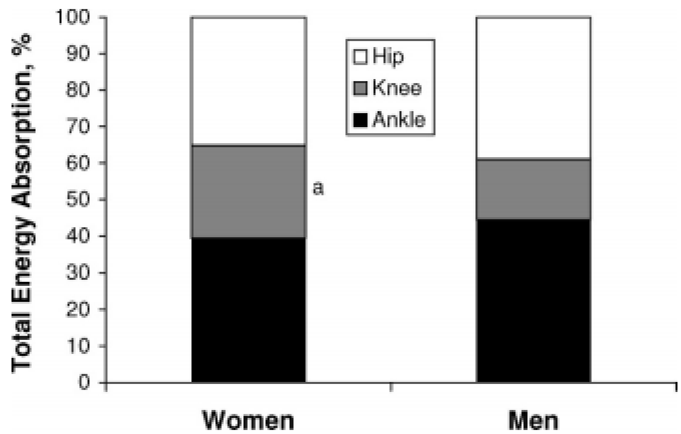


**Figure 7. Female and male relative energy absorption by joint.** [a]Energy absorption was greater in women than in men at the knee joint ($P$ < .001).

**Table. Mean ± SD Values, Effect Sizes, and 95% Confidence Intervals for Sagittal Hip, Knee, and Ankle Energy Absorption and Torsional Joint Stiffnesses**

| | Mean ± SD | | | 95% Confidence Interval | |
|---|---|---|---|---|---|
| | Women | Men | Effect Size | Women | Men |
| Energy absorption, J × $N^{-1}$ × $m^{-1}$ [a,b] | | | | | |
| Hip | 0.07 ± 0.03 | 0.08 ± 0.04 | 0.29 | 0.063–0.085 | 0.067–0.089 |
| Knee[c] | 0.05 ± 0.02 | 0.03 ± 0.01 | 1.16 | 0.045–0.056 | 0.024–0.036 |
| Ankle | 0.09 ± 0.03 | 0.09 ± 0.03 | 0.10 | 0.073–0.093 | 0.076–0.096 |
| Torsional stiffness, Nm × $N^{-1}$ × $m^{-1}$ × $degrees^{-1}$ [a,b] | | | | | |
| Hip[d] | 0.007 ± 0.004 | 0.010 ± 0.008 | 0.60 | 0.005–0.009 | 0.008–0.012 |
| Knee | 0.001 ± 0.0001 | 0.001 ± 0.0005 | 0.14 | 0.001–0.001 | 0.001–0.001 |
| Ankle | 0.002 ± 0.0004 | 0.002 ± 0.0006 | 0.06 | 0.002–0.002 | 0.002–0.003 |

[a] Body weight was measured in newtons (N).
[b] Height was measured in meters (m).
[c] Indicates energy absorption in the knee was greater in women than in men ($P$ = .017).
[d] Indicates torsional stiffness in the hip was greater in men than in women ($P$ = .003).

Compared with men, women were weaker in both the knee extensor (women: 2.28 ± 0.43 Nm/kg; men: 2.61 ± 0.41 Nm/kg; $P = .001$) and knee flexor (women: 1.78 ± 0.25 Nm/kg; men: 2.12 ± 0.28 Nm/kg; $P < .001$) muscle groups. These observed sex differences have been reported[18] in a large subset of the population, which was also included in this study. Because of these sex differences, we used stepwise regressions of energetics and stiffnesses within each sex subgroup, as well as within each joint. The regression for knee energy absorption in women revealed that greater knee extensor MVIC predicted greater knee energy absorption ($R^2 = 0.11$, $P = .04$), with knee flexor MVIC not explaining more variance ($P > .05$). The linear regression for hip torsional stiffness in women revealed that greater knee flexor MVIC predicted greater hip torsional stiffness ($R^2 = 0.12$, $P = .03$), with knee extensor MVIC not explaining more variance ($P > .05$). No other regressions revealed significant findings ($P > .05$).

## DISCUSSION

The main finding of our study was that during a countermovement task (the drop-jump landing cycle), women absorbed more energy about the knee than did men during the first landing phase of a drop-jump task. Additionally, men demonstrated greater hip joint stiffness than women. Furthermore, in women, greater knee extensor strength predicted greater knee energy absorption during this task, whereas greater knee flexor strength predicted greater hip torsional stiffness.

### Energetics and Stiffnesses

Our findings need to be viewed in light of the demands of the task performed. The first landing phase of the drop-jump task requires the body to arrest its downward momentum and then immediately propel the body upward. This task was chosen because Olsen et al[17] reported that acute joint injury occurs during tasks in which a sudden redirection of the center of mass occurs. Such a task is dominated by the stretch-shortening cycle (ie, concentric contraction phase is immediately preceded by an eccentric or active prestretch phase).[14]

We appear to be the first to report sex comparisons of energetics and torsional joint stiffnesses during a countermovement activity. At the knee joint, women absorbed 69% more energy about the knee joint than did men. This energy absorption is thought to be indicative of the eccentric muscle function during the loading process.[4,23] In addition, landing tasks that have been cognitively manipulated to be “soft” have been characterized by greater energy absorption compared with tasks that were cognitively manipulated to be “stiff.”[1,2,24] Given the large observed effect size for sex (1.16), the women in our study appeared to use a knee joint strategy during the stretch-shortening cycle that demonstrated the characteristics of greater energy absorption (soft). This is supported by the female and male knee flexion curves[18] shown in Figure 5 demonstrating that both sexes started in similar knee flexion angles but that women began to increase their excursion as the braking phase of landing continued. Furthermore, researchers[1] have revealed increased knee joint energy absorption during increasingly more challenging tasks. This increased energy absorption about the knee joint in women may result from the task being more challenging to them, as a similar drop height was used for all participants.

In a study using computer modeling, Sandler and Robinovitch[10] demonstrated that increasing energy absorption during a backward fall helped to decrease the severity of impact. Although the focus on falling is tangential to landing investigations, it is evidence that increasing energy absorption of the lower extremity is important to reducing the severity of an impact. Soft landing styles have been associated with tasks of increasing demand on the musculoskeletal system.[1] From this information, one could infer that the women in our study adopted an energy absorption pattern that was designed to minimize impact severity of a task that was relatively more difficult for women than for men. However, the role of greater or lesser energy absorption about the knee relative to the risk of ACL injury currently is not established in the literature.

As expected and previously demonstrated in a subset of the population we studied,[18] women demonstrated lower strength of the knee extensors and flexors when normalized to body size.[25,26] Given that energy absorption is thought to represent eccentric muscle action,[4] there seems to be a relationship between strength and energy absorption. Our data revealed that in women only, increasing knee extensor MVIC predicted a greater amount of knee energy absorption during the initial landing phase of a drop jump. Again, this may have occurred because it was a more demanding task for the women. In their study of a double-leg drop-landing task, Zhang et al[1] demonstrated that as the height of drop increased, more energy was absorbed by the joints of the lower extremity. Thus, women with increased knee extensor strength likely were better able to minimize the impact severity. Sandler and Robinovitch[10] reported that decreasing strength would decrease the effectiveness of energy absorption and, in turn, would increase impact severity. In addition, in a study of strength and landing biomechanics, Lephart et al[25] reported that female athletes had less thigh muscle strength, which they suggested was related to increased stiffness (as determined by less knee flexion excursion) upon landing; however, no correlational analyses were performed. They[25] suggested that a lack of strength negatively affected the manner in which female athletes attenuated the impact of landing. Collectively, it appears that increasing strength in women may be beneficial to controlling the body during landings. However, a large amount of the variance in knee energy absorption (89%) remains unexplained by thigh strength. Structural factors that differ between sexes (eg, joint laxity, lower extremity, alignment/posture) may also affect dynamic function.[27] Thus, it is unlikely that sex differences in landing mechanics are solely due to a single factor; it is more likely that multiple neuromuscular and structural factors interact to explain often-observed sex differences in landing mechanics.

The finding of greater hip torsional stiffness in men than women (moderate effect size of 0.60), along with the previously described differences in energy absorption, also indicated an overall more stiff landing in men. To test this suggestion, we used postanalysis testing for sex differences in overall leg spring stiffness (change in vertical ground reaction force/change in center of mass displacement). Simple 1-way ANOVA revealed no differences between women (6.9% ± 2.4% body weight $m^{-1}$) and men (6.7% ± 2.5% body weight $m^{-1}$) ($F_{1,79} = 0.237$, $P = .63$). Sagittal-plane torsional joint stiffnesses dictate the amount of flexion excursion in response to a given joint moment.[28] If hip stiffness increases with no additional change in torsional stiffness of the other joints, the overall stiffness of the system increases.[28] The reason for the opposite findings of increased hip stiffness in men and no sex differences in overall system stiffness may be that other factors contributed to overall system stiffness (energy absorption in the head-arms-trunk segment or energy dissipation through soft tissue vibration may have countered the increased hip stiffness in males).[29]

Although numerous factors can influence joint stiffness, joint stiffness apparently relies on the level of muscular activation around the joint.[28] One rationale for the increased hip torsional stiffness may be the task demands. In our investigation, participants were asked to drop down and then immediately jump as high as possible. In complex movements, such as drop jumping, "stiffening" of the system may permit more effective force transmission, thus allowing maximal performance.[30] This increased stiffness in men may be a mechanism for better transmission of force of the hip extensor muscles to the bony segments. The women in our investigation appeared to have used a less stiff strategy about the hip during the measured countermovement task. Researchers[3] have suggested that this observed decreased stiffness allows for excessive joint range of motion, potentially leading to soft tissue injury. This is supported by the female and male hip flexion curves in Figure 4, which demonstrate greater hip flexion in women as the landing task progresses.[18]

Our findings of greater knee energy absorption in women without subsequent sex differences in knee joint stiffness and of greater hip stiffness in men without subsequent sex differences in hip joint energy were somewhat contrary to those of previous work[1,24,31]indicating that stiffness and energy absorption are inversely related. To better compare current data with those of previous work, we conducted a secondary Pearson product moment correlation analysis. It revealed inverse relationships between hip energy absorption and hip stiffness (women: $r = -0.43$, $P = .005$; men: $r = -0.57$, $P < .001$) and knee energy absorption and knee stiffness (women: $r = -0.57$, $P = .005$; men: $r = -0.56$, $P < .001$). Although previous researchers[31] have reported that knee work absorption and overall leg impedance were highly correlated ($r = 0.74$) in drop landings, the more moderate relationships that we found may exist because of task differences (landing only in previous work) or differences in the specific type of stiffness data with which the energetics were correlated. Still, it appears that our study supports the well-accepted notion that stiffer landings are characterized by less energy absorption about the lower extremity joints.

Greater knee flexor strength may be able to predict greater hip torsional stiffness in women because the hamstrings comprise a 2-joint muscle that provides for knee flexion and hip extension. Thus, women who had greater hamstrings strength were able to maintain a higher level of torsional stiffness about the hip joint. This relationship was not found in men, so men, who on average were stronger, may have possessed a critical amount of strength to perform the task with increased stiffness about the hip. This increased stiffness in women may be a byproduct of increased co-contraction about the knee joint (this could theoretically be associated with a stiffer joint), which is unlikely as a result of the observed greater energy absorption about the knee in women. A more likely rationale is that women were using hamstrings muscle function to help control hip joint motion and resulting trunk position. Previous work[32] in falling has demonstrated that the hamstrings likely play a role in helping to control position of the trunk segment. As is the case with knee energy absorption, the large amount of variance in hip torsional stiffness remained unexplained after accounting for thigh strength.

## Limitations

We acknowledge limitations of the research design. Energy absorption was calculated as the integral of the entire negative power curve from foot contact to peak negative center-of-mass displacement and, thus, may limit direct comparison with previous research,[8]in which the integral of the power curve during the first 100 milliseconds after foot contact was calculated. We decided to use the entire flexion motion because we wanted to best represent all energy absorption of the extensor muscles during the initial landing phase of the drop-jump task. Although this method of quantifying energy absorption has been used previously[2,4] and has been thought to represent energy absorption of the muscles, we acknowledge that the specific contributions of active and passive structures about the joint to total energy absorption cannot be differentiated. Furthermore, the summed hip, knee, and ankle energy absorption accounted for only 78.1% ± 16.4% and 79.1% ± 17.5% of the potential energy associated with the 0.45-m drop in men and women, respectively. Researchers should investigate other sources of energy absorption in the system, such as energy absorption in other planes of lower extremity motion, energy absorption in the head-arms-trunk segment, or energy dissipation through soft tissue vibration.[29]

Including electromyographic data also would help in understanding contractile tissue function during the task by providing knowledge of relative activation levels, thus allowing a better understanding of what percentage of maximal activation was used to complete the energy absorption phase of the task. Future investigators also need to study hip and ankle strength to determine their influences on landing energetics and torsional joint stiffnesses.

The ecological (context) validity of the findings may be lessened because participants performed tasks barefooted. This was done to help control for the effects of various types of footwear across participants. Future researchers should determine if uniform footwear would alter the results of the investigation.

### Clinical Implications

We are the first to look at the relationships between thigh strength and lower extremity sagittal-plane energetics and stiffness. The observed sex differences of greater knee energy absorption and decreased hip torsional stiffness in women were explained, in part, by knee extensor and knee flexor strength. By functionally changing thigh muscle strength through training programs, it may be possible to influence energy absorption patterns of the extensor muscles during a task in which a rapid deceleration occurs, followed by a task that redirects the bodily position. Thus, this finding may lend support to the inclusion of strength components in injury prevention programs. However, large amounts of the variance in knee energy absorption (89%) and hip torsional stiffness (88%) remain unexplained by thigh strength, indicating that other factors also contribute to these sex differences. To build on the findings of our work, future prospective work is needed to determine if the observed sex-specific energy absorption and torsional stiffness patterns are predictive of ACL injury. In additional future work, researchers also should look at other predictive factors of the observed sex differences in landing biomechanics to most efficiently design and implement injury-prevention programs.

## ACKNOWLEDGMENTS

This study was supported by research grant R01 AR053172 from the National Institutes of Health and the National Institute of Arthritis and Musculoskeletal and Skin Diseases.

## REFERENCES


1. Zhang, S. N. , B. T. Bates , and J. S. Dufek . Contributions of lower extremity joints to energy dissipation during landings. *Med Sci Sports Exerc* 2000. 32 4:812–819.

2. Devita, P. and W. A. Skelly . Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Med Sci Sports Exerc* 1992. 24 1:108–115.

3. Butler, R. J. , H. P. Crowell III , and I. M. Davis . Lower extremity stiffness: implications for performance and injury. *Clin Biomech (Bristol, Avon)* 2003. 18 6:511–517.

4. McNitt-Gray, J. L. Kinetics of the lower extremities during drop landings from three heights. *J Biomech* 1993. 26 9:1037–1046.

5. Farley, C. T. and D. C. Morgenroth . Leg stiffness primarily depends on ankle stiffness during human hopping. *J Biomech* 1999. 32 3:267–273.

6. Arampatzis, A. , G. P. Bruggemann , and V. Metzler . The effect of speed on leg stiffness and joint kinetics in human running. *J Biomech* 1999. 32 12:1349–1353.

7. Granata, K. P. , S. E. Wilson , and D. A. Padua . Gender differences in active musculoskeletal stiffness: part I. Quantification in controlled measurements of knee joint dynamics. *J Electromyogr Kinesiol* 2002. 12 2:119–126.

8. Decker, M. J. , M. R. Torry , D. J. Wyland , W. I. Sterett , and R. J. Steadman . Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clin Biomech (Bristol, Avon)* 2003. 18 7:662–669.

9. Schmitz, R. J. , A. S. Kulas , D. H. Perrin , B. L. Riemann , and S. J. Shultz . Sex differences in lower extremity biomechanics during single leg landings. *Clin Biomech (Bristol, Avon)* 2007. 22 6:681–688.

10. Sandler, R. and S. Robinovitch . An analysis of the effect of lower extremity strength on impact severity during a backward fall. *J Biomech Eng* 2001. 123 6:590–598.

11. Shultz, S. J. , R. J. Schmitz , and A. D. Nguyen . Research Retreat IV: ACL injuries—the gender bias: April 3–5, 2008, Greensboro, NC. *J Athl Train* 2008. 43 5:530–531.

12. Mizner, R. L. , J. K. Kawaguchi , and T. L. Chmielewski . Muscle strength in the lower extremity does not predict postinstruction improvements in the landing patterns of female athletes. *J Orthop Sports Phys Ther* 2008. 38 6:353–361.

13. Santello, M. Review of motor control mechanisms underlying impact absorption from falls. *Gait Posture* 2005. 21 1:85–94.

14. Moran, K. A. and E. S. Wallace . Eccentric loading and range of knee joint motion effects on performance enhancement in vertical jumping. *Hum Movement Sci* 2007. 26 6:824–840.

15. Kovacs, I. , J. Tihanyi , P. Devita , L. Racz , J. Barrier , and T. Hortobagyi . Foot placement modifies kinematics and kinetics during drop jumping. *Med Sci Sports Exerc* 1999. 31 5:708–716.

16. McCaulley, G. O. , P. Cormie , M. J. Cavill , J. L. Nuzzo , Z. G. Urbiztondo , and J. M. McBride . Mechanical efficiency during repetitive vertical jumping. *Eur J Appl Physiol* 2007. 101 1:115–123.

17. Olsen, O. E. , G. Myklebust , L. Engebretsen , and R. Bahr . Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. *Am J Sports Med* 2004. 32 4:1002–1012.

18. Shultz, S. J. , A. D. Nguyen , M. D. Leonard , and R. J. Schmitz . Thigh strength and activation as predictors of knee biomechanics during a drop jump task. *Med Sci Sports Exerc* 2009. 41 4:857–866.

19. Leardini, A. , A. Cappozzo , F. Catani , et al. Validation of a functional method for the estimation of hip joint centre location. *J Biomech* 1999. 32 1:99–103.

20. Madigan, M. L. and P. E. Pidcoe . Changes in landing biomechanics during a fatiguing landing activity. *J Electromyogr Kinesiol*2003. 13 5:491–498.

21. Kadaba, M. P. , H. K. Ramakrishnan , M. E. Wootten , J. Gainey , G. Gorton , and G. V. Cochran . Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. *J Orthop Res* 1989. 7 6:849–860.

22. Gagnon, D. and M. Gagnon . The influence of dynamic factors on triaxial net muscular moments at the L5/S1 joint during asymmetrical lifting and lowering. *J Biomech* 1992. 25 8:891–901.

23. McNitt-Gray, J. L. , D. M. E. Hester , W. Mathiyakom , and B. A. Munkasy . Mechanical demand and multijoint control during landing depend on orientation of the body segments relative to the reaction force. *J Biomech* 2001. 34 11:1471–1482.

24. Kulas, A. S. , T. C. Windley , and R. J. Schmitz . Effects of abdominal postures on lower extremity energetics during single-leg landings. *J Sport Rehabil* 2005. 14 1:58–71.

25. Lephart, S. M. , C. M. Ferris , B. L. Riemann , J. B. Myers , and F. H. Fu . Gender differences in strength and lower extremity kinematics during landing. *Clin Orthop Relat Res* 2002. 401:162–169.

26. Uhorchak, J. M. , C. R. Scoville , G. N. Williams , R. A. Arciero , P. St Pierre , and D. C. Taylor . Risk factors associated with noncontact injury of the anterior cruciate ligament: a prospective four-year evaluation of 859 West Point cadets. *Am J Sports Med* 2003. 31 6:831–842.

27. Nguyen, A. D. and S. J. Shultz . Sex differences in clinical measures of lower extremity alignment. *J Orthop Sports Phys Ther*2007. 37 7:389–398.

28. Farley, C. T. , H. H. P. Houdijk , C. V. Strien , and M. Louie . Mechanism of leg stiffness adjustment for hopping on surfaces of different stiffnesses. *J Appl Physiol* 1998. 85 3:1044–1055.

29. Self, B. P. and D. Paine . Ankle biomechanics during four landing techniques. *Med Sci Sports Exerc* 2001. 33 8:1338–1344.

30. Bojsen-Moller, J. , S. P. Magnusson , L. R. Rasmussen , M. Kjaer , and P. Aagaard . Muscle performance during maximal isometric and dynamic contractions is influenced by the stiffness of the tendinous structures. *J Appl Physiol* 2005. 99 3:986–994.

31. Kulas, A. S. , R. J. Schmitz , S. J. Schultz , M. A. Watson , and D. H. Perrin . Energy absorption as a predictor of leg impedance in highly trained females. *J Appl Biomech* 2006. 22 3:177–185.

32. Pijnappels, M. , M. F. Bobbert , and J. H. van Dieen . How early reactions in the support limb contribute to balance recovery after tripping. *J Biomech* 2005. 38 3:627–634.